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Comparison of the Windkessel model and structured-tree model applied to prescribe outflow boundary conditions for a one-dimensional arterial tree model

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ABSTRACT

One-dimensional (1D) modeling is a widely adopted approach for studying wave propagation phenomena in the arterial system. Despite the frequent use of the Windkessel (WK) model to prescribe outflow boundary conditions for 1D arterial tree models, it remains unclear to what extent the inherent limitation of the WK model in describing wave propagation in distal vasculatures affect hemodynamic variables simulated at the arterial level. In the present study, a 1D model of the arterial tree was coupled respectively with a WK boundary model and a structured-tree (ST) boundary model, yielding two types of arterial tree models. The effective resistances, compliances and inductances of the WK and ST boundary models were matched to facilitate quantitative comparisons. Obtained results showed that pressure/flow waves simulated by the two models were comparable in the aorta, whereas, their discrepancies increased towards the periphery. Wave analysis revealed that the differences in reflected waves generated by the boundary models were the major sources of pressure wave discrepancies observed in large arteries. Additional simulations performed under aging conditions demonstrated that arterial stiffening with age enlarged the discrepancies, but with the effects being partly counteracted by physiological aortic dilatation with age. These findings suggest that the method adopted for modeling the outflow boundary conditions has considerable influence on the performance of a 1D arterial tree model, with the extent of influence varying with the properties of the arterial system.

1. Introduction

One-dimensional (1D) models have been widely employed as a useful tool for studying wave propagation phenomena in the arterial system (Stergiopulos et al., 1992; Sherwin et al., 2003; Liang et al., 2009a; Müller and Toro, 2014; Mynard and Smolich, 2015). In most of the cases, 1D modeling is limited to large arteries, which requires the definition of outflow boundary conditions so that wave reflections at the distal ends of peripheral arteries can be captured. The Windkessel (WK) model has been frequently used to prescribe outflow boundary conditions for 1D arterial tree models (Alastruey et al., 2008; Stergiopulos et al., 1992; Reymond et al., 2009). Physiologically, the WK model is defective because it overlooks the detailed anatomical and mechanical properties of the vascular system being modeled (Shi et al., 2011; Westerhof et al., 2009), thus compromising its capability for describing wave propagation phenomena. When the model is applied to represent the vascular systems distal to an arterial tree, the limitation may propagate upstream to compromise the fidelity of simulated pressure/flow waves in large arteries. By contrast, the structured-tree (ST) model provides a more sophisticated means to characterize the dynamic pressure–flow relationship in a distal vascular system with multiple generations of bifurcation (Olufsen, 1999; Olufsen et al., 2000; Cousins et al., 2013; Cousins and Gremaud, 2014). Advanced applications of ST-like models have been reported as well (Perdikaris et al., 2015). The ST model, given the inherent assumptions, is by far a simplified representation of the actual vascular anatomy which is usually organ/tissue-specific (Blanco et al., 2014, 2015), nevertheless, it provides a relatively reasonable approach to modeling smaller distal vessels whose anatomical parameters are not explicitly known, such as arterioles.

Theoretically, the differences of the WK and ST models can be assessed by comparing the impedances resulting from the models.

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However, it remains unclear how the differences in impedance affect the characteristics of pressure/flow waves simulated at the arterial level. In the present study, a 1D model of the arterial tree was coupled respectively with a WK outflow boundary model and a ST outflow boundary model to construct two types of arterial tree model (namely, 1D-WK model and 1D-ST model). To permit quantitative comparisons, in addition to uniform parameter assignment for the 1D model, parameters used in the WK and ST models were tuned in such a way that the effective resistance, compliance and inductance were equivalent for the two models. Moreover, the two arterial tree models were compared in the context of aging to investigate whether the sensitivity of simulated pressure/flow waves to outflow boundary models is affected by the properties of large arteries.

2. Methods

1D modeling was applied to the 55 largest arteries (Liang et al., 2009a, 2009b). The distal end of each peripheral artery was supported by a WK or a ST outflow boundary model that represents the distal vascular systems (see Fig. 1). The entrance of the ascending aorta was connected to a lumped-parameter model of the left heart so that the resulting model can account for ventricular–arterial coupling and spontaneously capture hemodynamic changes resulting from the application of different outflow boundary models.

2.1. Governing equations of the 1D model

1D governing equations for blood flows in an artery can be obtained by integrating the three-dimensional Navier–Stokes equations over the cross-section of the artery (Formaggia et al., 2006). Obtained 1D hemodynamic equations are expressed as (Liang et al., 2009b)

$$\frac{\partial A}{\partial t}+\frac{\partial Q}{\partial z}=0, \tag{1}$$

$$\frac{\partial Q}{\partial t}+\frac{\partial}{\partial z}\left(\alpha\frac{Q^2}{A}\right)+\frac{A}{\rho}\frac{\partial P}{\partial z}+F_r\frac{Q}{A}=0, \tag{2}$$

where t is the time and z the axial coordinate; A, Q and P represent the cross-sectional area, volume flux and pressure, respectively; α is the momentum-flux correction coefficient and F_r the friction force per unit length. Herein, a Poiseuille cross-sectional velocity profile was assumed, based on which, α, F_r were calculated to be 4/3, $-8\pi\upsilon$ (υ is the kinematic viscosity), respectively.

The system of Eqs. (1) and (2) was completed by a constitutive equation that accounts for the viscoelastic deformation of arterial wall (Bessems et al., 2008; Liang et al., 2013).

$$P+\tau_\sigma\frac{\partial P}{\partial t}=\phi(A)+\tau_\varepsilon\frac{\partial\phi(A)}{\partial t}, \text{with } \phi(A)=\frac{Eh}{r_0(1-\sigma^2)}\left(\sqrt{\frac{A}{A_0}}-1\right)+P_0 \tag{3}$$

Here, τ_σ and τ_ε refer respectively to the relaxation times for constant stress and strain; P_0 is the reference pressure; E is the Young's modulus; h is the wall thickness; σ is the Poisson's ratio, here taken to be 0.5; A_0 and r_0 represent the cross-sectional area and radius of artery at the reference pressure, respectively.

The 1D hemodynamic equations of each artery were linked to those of the neighboring arteries by imposing continuity of mass flux and total pressure at the bifurcations so that blood flows can be solved for the entire arterial tree (Liang et al., 2009a, 2009b; Mynard and Nithiarasu, 2008).

2.2. Modeling of the left heart

The classical elastance model was employed to account for the pumping function of the left heart (including the left atrium and ventricle) (see Fig. 1). The mathematical expressions followed from those used in our previous studies (Liang et al., 2014b). The mitral and aortic valves were modeled in such way that no reversed flows are allowed to occur. For more details on the heart model, please refer to our previous studies (Liang et al., 2014a, 2014b).

2.3. Modeling of the outflow boundary conditions of the artery tree

Vasculatures distal to the peripheral arteries are featured by complex anatomical structures and heterogeneous mechanical properties. Anatomically, the distal vasculatures can be categorized into distal arteries, arterioles, capillaries, venules and veins, among which distal arteries and arterioles are the major determinant of the wave reflection conditions at the distal ends of peripheral arteries and were herein modeled using a WK model and a ST model, respectively.

2.3.1. Windkessel (WK) model

The WK model consists of four elements, namely, two resistors (R_1, R_2), a capacitor (C) and an inductor (L) (see Fig. 1). These elements correspond to the lumped properties of the distal vasculatures, including the viscous resistance, the distensibility of vascular walls and the inertia of blood. The input impedance (Z) of the model can be expressed in the frequency domain as

$$Z(\omega)=R_1+\frac{R_2+i\omega L}{1+iR_2\omega C-\omega^2LC}, \tag{4}$$

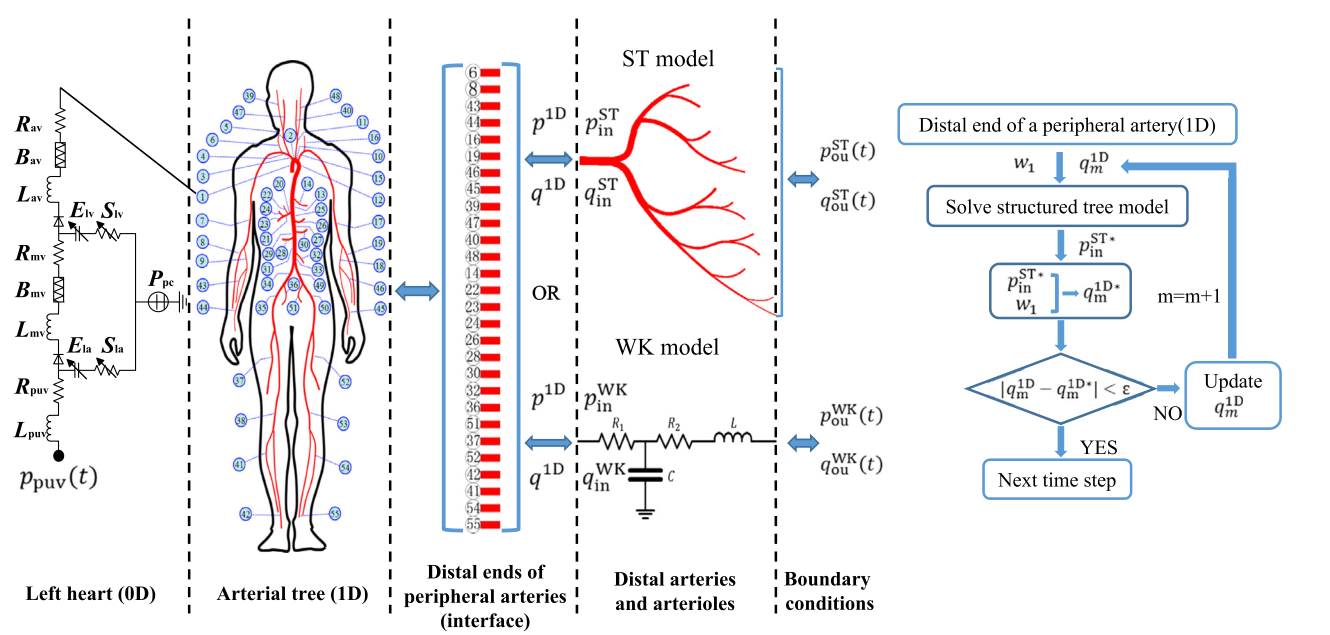

Fig. 1. Schematic description of the arterial tree model (left) and the flowchart of the numerical procedure for coupling the 1D and ST models (right). The entrance of the artery tree is coupled with a lumped model of the left heart, while the distal ends of the tree are coupled with WK or ST boundary models. Notations of parameters used in the heart model: E, elastance; L, inductance; R, viscous resistance; C, compliance; B, Bernoulli's resistance; S, viscoelasticity coefficient of myocardial wall; P_{pc}, pericardium pressure; p_{puv}, pulmonary venous pressure. Subscripts: la, left atrium; lv, left ventricle; mv, mitral valve; av, aortic valve. For notations of parameters used in the flowchart, see the text for details.